Klinik für Pferdechirurgie

Departement für Pferde

der Vetsuisse-Fakultät Universität Zürich

Klinikdirektion: Prof. Dr. med. vet., FVH, Anton Fürst, Dipl. ECVS

Arbeit unter wissenschaftlicher Betreuung von

Dr. med. vet., FVH, Michelle Jackson, Dipl. ECVS

The influence of aluminium, steel and polyurethane shoeing systems and of the unshod hoof on the injury risk of a horse kick – An ex vivo experimental study

Inaugural-Dissertation

zur Erlangung der Doktorwürde der

Vetsuisse-Fakultät der Universität Zürich

vorgelegt von

Miriam Maria Flurina Sprick

Tierärztin

von Chur, Graubünden

genehmigt auf Antrag von

Prof. Dr. med. vet. Anton Fürst, Referent

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Für meine Mutter und meinen Vater, die mir beide zu ihren Teilen die Liebe zu Tier und Natur eröffnet und mir all dies ermöglicht haben.

Inhaltsverzeichnis

Vetsuisse-Fakultät Universität Zürich 2017

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The influence of aluminium, steel and polyurethane shoeing systems and of the unshod hoof on the injury risk of a horse kick – An ex vivo experimental study

Summary:

Objectives: To evaluate the damage inflicted by unshod hoof and the horseshoe materials (steel, aluminium and polyurethane (PU)) on the long bones of horses after a simulated kick. Methods: 64 equine radii and tibiae were evaluated using a drop impact test setup. An impactor with a steel, aluminium, PU or hoof horn head was dropped onto prepared bones. An impactor velocity of 8 m/s was used with all four materials and a velocity of 12 m/s also was used with the PU and hoof horn heads. The impact process was analysed using a high-speed camera and physical parameters including peak contact force and impact duration were calculated. Results: The probability of a fracture was 75 % for steel and 81 % for aluminium, whereas PU and hoof horn did not damage the bones at 8 m/s. At 12 m/s, the probability of a fracture was 25 % for PU and 12.5 % for hoof horn. The peak contact force and impact duration differed significantly between ‘hard materials’ (aluminium and steel) and ‘soft materials’ (PU and hoof horn). Clinical Significance: The observed bone injuries were similar to those seen in analogous experimental studies carried out previously and comparable to clinical fracture cases suggesting that the simulated kick was realistic. The probability of fracture was significantly higher for steel and aluminium than for PU and hoof horn, which suggests that the horseshoe material has a significant influence on the risk of injury in humans or in horses kicked by a horse.

Keywords: Horse, shoeing, impact load, kick injury

Vetsuisse-Fakultät Universität Zürich 2017

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Der Einfluss von Aluminium, Stahl und Polyurethane Beschlagssystemen und des unbeschlagenen Hufes auf das Verletzungsrisiko durch einen Hufschlag – Eine experimentelle ex vivo Studie

Zusammenfassung:

Ziel: Experimentelle Untersuchung des Einflusses der Hufbeschläge Aluminium, Stahl und Polyurethane (PU) und unbeschlagener Hufe auf das Verletzungsrisiko langer Röhrenknochen durch einen Hufschlag. Methoden: In einem Fallturm wurden Schlagkörper mit Köpfen aus Aluminium, Stahl, PU oder Horn mit einer Geschwindigkeit von 8m/s auf 64 präparierte equine Radii und Tibiae fallen gelassen. Für die Materialien PU und Horn wurde das Experiment mit 12m/s wiederholt. Der Aufschlag wurde mit einer Hochgeschwindigkeits-kamera gefilmt um die physikalischen Parameter maximale Kontaktkraft und Aufschlagsdauer zu kalkulieren. Resultate: Die Schadenswahrscheinlichkeit bei 8m/s betrug 75% bei Stahl und 81% bei Aluminium. Bei Schlägen mit PU und Horn traten keine Schäden auf. Die Schadenswahrscheinlichkeit bei 12m/s betrug 25% bei PU und 12.5% bei Horn. Die maximale Kontaktkraft und die Aufschlagsdauer wiesen signifikante Unterschiede zwischen harten Materialien (Aluminium und Stahl) und weichen Materialien (PU und Horn) auf. Klinische Bedeutung: Die erhaltenen Knochenschäden entsprachen sowohl ähnlichen Studien als auch klinischen Fällen, weshalb von einer realistischen Simulation eines Hufschlages ausgegangen wird. Die Frakturwahrscheinlichkeit durch Schläge mit Aluminium oder Stahl war signifikant höher als mit PU und Horn. Der Hufbeschlag übt also einen signifikanten Einfluss auf das Verletzungsrisiko von Menschen oder Pferden bei einem Hufschlag aus.

Schlüsselwörter: Pferd, Hufbeschlag, Stossbelastung, Schlagverletzung

Introduction

The structure and material properties of long bones such as radii and tibiae are not well suited for resisting side impact forces (1). Therefore, loads applied to long bones in a direction perpendicular to the longitudinal axis often result in serious damage including fissures and fractures (2). Long bone fractures are common in pedestrians hit by vehicles (3) and in people kicked by a horse (4, 5). The most common cause of long bone fractures in horses is a kick from another horse (6, 7). A retrospective study found that approximately 47 % of equine injuries caused by a kick from another horse were bone fissures or fractures. Thus, a fractured radius or tibia as a result of a kick from another horse is common (4).

It was shown that threatening to kick is the most common sign of aggression in herds of wild horses (8). In Switzerland, many horses are kept on small pastures because of limited space, which contributes to aggressive behaviour among herd mates (4). Group housing systems have become very popular (4, 9), but prevention of kick injuries among horses is critical. In these systems shoeing with steel or aluminium shoes is often prohibited whereas the use of alternative shoeing systems made of soft materials, such as polyurethane (PU), is commonly accepted. It is generally assumed that a kick delivered by an unshod hoof reduces the likelihood of severe injury in horses or humans; however, there is no scientific evidence for this. Likewise, the effect of PU horseshoes on the severity of a kick injury has not been investigated. The primary function of a horseshoe is to protect the hoof against wear or damage during movement. Horseshoes are traditionally made of steel because steel is a high-strength material and offers excellent wear protection. A disadvantage is that the additional weight and stiffness compared with hoof horn may limit the shock-absorbing features of the hoof. Horseshoes made of aluminium, a less dense metal than steel, were introduced mainly to reduce inertia forces during racing. Compared with steel horseshoes, those made of PU have a lower resistance to wear but provide better damping and shock absorption (10, 11). Furthermore, they significantly reduce craniocaudal deceleration forces after a step impact, resulting in less jarring of the distal limb (12). The effect of a simulated horse kick on equine long bones has been investigated and evaluated for steel horseshoes (13, 14). To the authors' knowledge, the effect of other horseshoe materials as well as the unshod hoof on the severity of kick injuries has not been studied.

In the present experimental study, the physical conditions of a kick by a horse on equine long bones (tibia and radius) were simulated using four different materials: steel, aluminium, PU and hoof horn. The primary goal was to compare the potential of different horseshoe materials and of unshod hoof to cause a fracture in the event of a kick. Our primary hypothesis was that a kick delivered by an unshod hoof or a hoof with a PU shoe has a significantly lower probability of causing a long bone fracture than a kick delivered by a hoof shod with a steel or aluminium shoe. The secondary hypothesis was that a kick by an unshod hoof has a lower probability of causing a fracture than a kick delivered by a PU horseshoe. To explain the differences in fracture probability, physical parameters that characterise the impact process, such as peak contact force and impact duration, were measured.

Materials and methods

A total of 32 radii and 32 tibiae were collected from 19 horses of various breeds euthanized at the Clinic of Equine Surgery, University of Zurich, for various reasons between July 2001 and March 2016. The horses ranged in age from 4 to 26 years and included 15 geldings, one stallion and three mares. None of the horses had a history of chronic generalised bone disease or other disease affecting bone. Within 24 hours after euthanasia, the limbs were detached from the body and the soft tissues were removed from the bones. The periosteum was removed from the diaphysis in the region where the impactor would contact the bone. This was done to reduce any mitigating effect of the soft tissues and to create a "worst case" impact scenario. The bones were prepared similarly as described in (13, 14). The length of the bones varied between 36.2 and 43 cm. If needed the bones were shortened by equal amounts on both ends to a final length of 38 cm. The proximal and distal ends of the specimens were casted in polyurethane[a] up to 1cm above the metaphysis (Figure 1). The bones were wrapped in a cloth soaked in 0.9 % saline solution and placed in plastic bags in order to prevent drying. The bones used in this study had been stored at -20 °C for 15 years (n = 4), 11 years (n = 56) or two month (n = 4). Testing was done at room temperature.

Fig. 1: Dissected left tibia with the proximal and distal ends embedded in polyurethane. The blue cross indicates where the impactor was intended to impact the bone.

The test setup was the same as in previous studies (13, 14) with the exception of the impactor, which had an exchangeable head (Figure 2). The cylindrical impactor heads (length = 4.5cm, diameter = 2cm) were made of steel (S), aluminium (A), polyurethane (PU) and hoof horn (H) (Figure 3, Table 1 (15, 16)). A cylindrical steel core with a diameter of 6 mm was inserted into the PU cylinder to simulate a commercially available polyurethane horseshoe[b]. The horn cylinder was cut from the dorsal hoof wall of a hind foot. The different masses of the cylinders were balanced with additional weights attached to the aluminium body of the impactor to obtain a uniform total impactor mass of 2.00 kg. The same dropping facility described in previous studies (13, 14) was used to accelerate the impactor. The bones were fixed horizontally between two metal blocks with the medial side of the bone uppermost. The impactor was guided on a rail oriented perpendicular to and centred on the bone. An axial force of 2.4 kN was applied to the bone to simulate the load on the radius or tibia in the weight-bearing limb of a standing horse (13, 14).

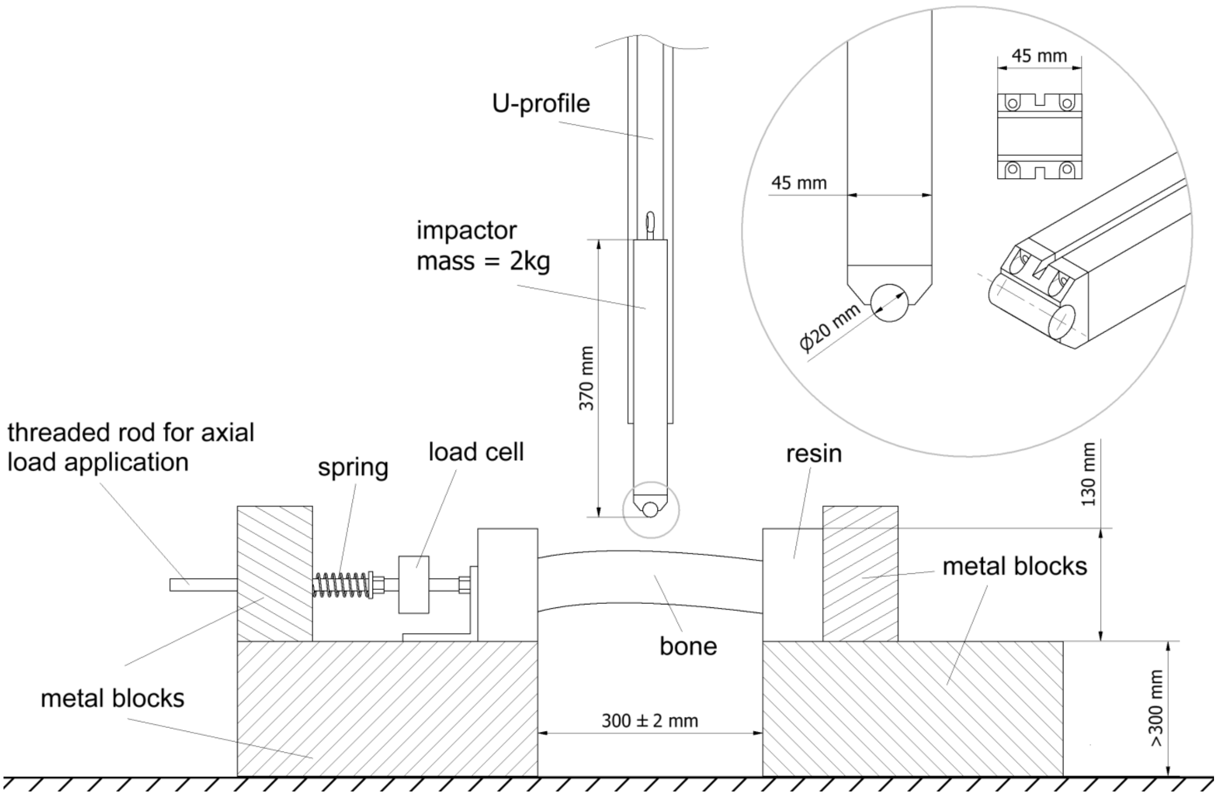

Fig. 2: Illustration of the test setup.

Fig. 3: Impactor heads: cylindrical steel, aluminium, polyurethane and hoof horn bars (from left to right).

	Steel	Aluminium	Polyurethan	Horn
Mat-Code / Source	S275	AlMgSi1	PUR 90	Hoof wall
Density [kg/m^3]	7‘873	2‘690	1‘254	1‘230
Elastic modulus [MPa]	210‘000*	68‘200*	294*	750

* Data from the literature (15, 16)

Table 1: Identification codes and mechanical properties of the four materials used. The elastic modulus of hoof horn was measured on cylinders (L = 18 mm, D = 8 mm) in compression load.

The 64 bones were divided into four groups of equal size to obtain a uniform distribution with respect to age, gender and type of bone (radius/tibia, left/right). A group did not contain more than one bone of the same horse to avoid bias caused by individual parameters such as bone strength. The groups were randomly assigned to one impactor head. The impact velocity was controlled by adjusting the impactor drop height. To facilitate testing of the hypotheses, the velocity at impact was set to a value that resulted in a high probability of fracture when the bone was impacted by the steel head. Based on a previous study (13), a velocity of 8 m/s (meters per second) fulfilled this requirement with a reported fracture probability of approximately 80 %.

In the first test series, which involved 64 simulated kick experiments, the impact velocity was set at 8 m/s. Each bone of each group was hit once.

To test the second hypothesis, it was necessary to increase the velocity to 12 m/s, since at 8 m/s unshod and PU shod hoof simulations did not result in bone fractures at all. A total of 32 simulated kick experiments were carried out and each of the bones in the hoof horn and PU groups were hit a second time.

After the impact tests, the bones were visually inspected for damage and photographed using a digital camera[c]. To evaluate fracture patterns, all bones with obvious signs of damage were examined radiographically in orthogonal projections (dorso-palmar/plantar and latero-medial[d]). All other bones were examined by computed tomography (40-slice scanner CT[e]). Settings included 120 kV, 100 mAs, 1 s tube rotation, a pitch of 0.65, 2 mm slice collimation with an increment of 2 mm reconstructed to 0.75 mm images applying a medium-frequency image reconstruction algorithm (soft tissue) and a high-frequency image reconstruction algorithm (bone), respectively. All bones were classified as fractured or not fractured, and incomplete (e.g. fissure) and complete fractures were differentiated. Complete fractures were divided into simple transverse, simple oblique, simple longitudinal, butterfly and comminuted fractures.

The peak contact force (F_{peak}) and the duration of the impact event, i.e., the time from when the impactor first contacted the bone to reversal of the impactor (t_{impact}), were determined to analyse the impact process using a high-speed video camera[f] with a rate of f = 40’000 frames per second. The selected frame rate provided a time resolution of $\Delta t = 1/f = 25$ μs at an optical resolution of 256×352 pixels (Figures 4a-c). The position of the impactor over time, y(t), and the fracture propagation across the bone were evaluated using computer-aided video tracking[g] with an accuracy of $\Delta y = \pm 0.01$mm. The impactor velocity, v(t), was calculated by numerical differentiation of the displacement function relative to time.

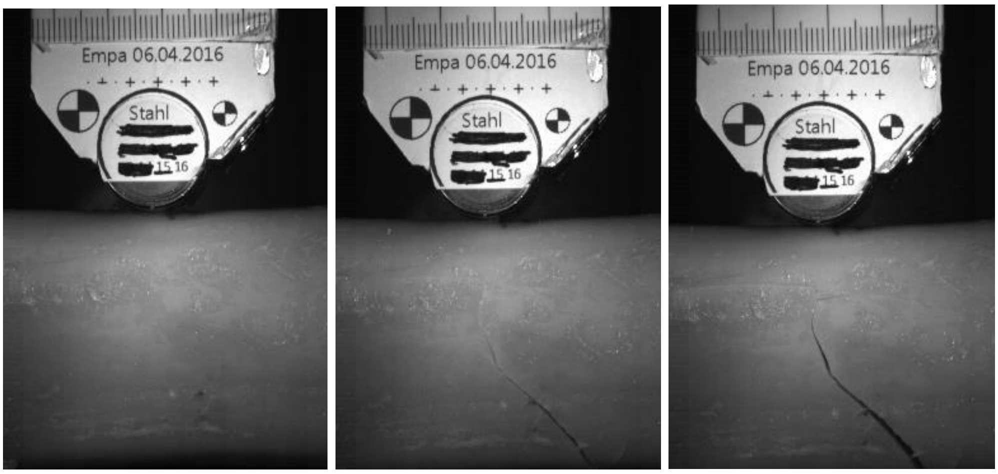

Fig. 4a-c: Three typical consecutive images of a high-speed video showing a long bone impacted with a steel impactor head. The images are ten frames apart ($\Delta t = 0.25$ ms), and the first image shows the point in time when the impactor first contacted the bone.

The contact force between the impactor and the bone, F(t), was calculated based on Newton's law: $F(t) = m \cdot a(t)$, where m is the impactor's mass and a(t) is the deceleration of the impactor as a function of time (derived from the numerical differentiation of the velocity v(t)). The time of first contact was considered to be when movement of the bone was first observed, whereas the time of reversal of the impactor was based on the measured velocity curve at v(t) = 0. The duration of the impact event was calculated from these measurements.

Statistics

Because of similar material properties, and due to the small number of undamaged bones, the materials steel and aluminium were combined into one group, named 'hard materials'. PU and horn were combined into one group named 'soft materials'. The software R[h] version 3.3.1 was used for statistical analysis. The difference in probability of fracture between hard and soft materials was analysed using Fisher's exact test. Regression analysis (linear models) was used to assess the effect of the test material on the outcome variables peak contact force (F_{peak}) and duration of the impact (t_{impact}). Type of bone (tibia or radius), side (left or right leg) and interaction effect between type of bone and material were included as potential explanatory variables. Model selection was based on AIC (Akaike information criterion) with lower values indicating a better model fit. Residuals were assessed visually for normality and homogeneity of variance. Tukey HSD 95% confidence interval (CI) was used to adjust for multiple comparisons between the different materials.

Results

Fracture probability

Impact at 8 m/s

Hard materials (steel and aluminium): Of the 16 bones impacted by steel, four had no detectable damage, macroscopically as well as under CT examination, and 12 had a fracture: three bones had a fissure and nine bones had a complete fracture (one simple transverse fracture, six simple oblique fractures, one simple longitudinal fracture and one comminuted fracture). The fracture probability was 75 % with a 95 % confidence interval [48 % to 93 %]. Of the 16 bones impacted by the aluminium, three remained undamaged (macroscopically and under CT examination) and 13 had a fracture: three bones had a fissure and ten bones had a complete fracture. There was one simple transverse fracture, three simple oblique fractures, one simple longitudinal fracture, three butterfly fractures (Figure 5) and two comminuted fractures. The fracture probability was 81 % with a 95 % CI [54 % to 96 %]. The probability of fracture of bones impacted by steel and aluminium were largely overlapping.

Soft materials (PU and hoof horn): The 32 bones impacted by PU or horn remained intact and fissures that were not visible macroscopically were ruled out in all bones using CT examination.

The fracture probability of bones impacted by hard and soft materials differed significantly ($p < 0.001$, Fisher's exact test).

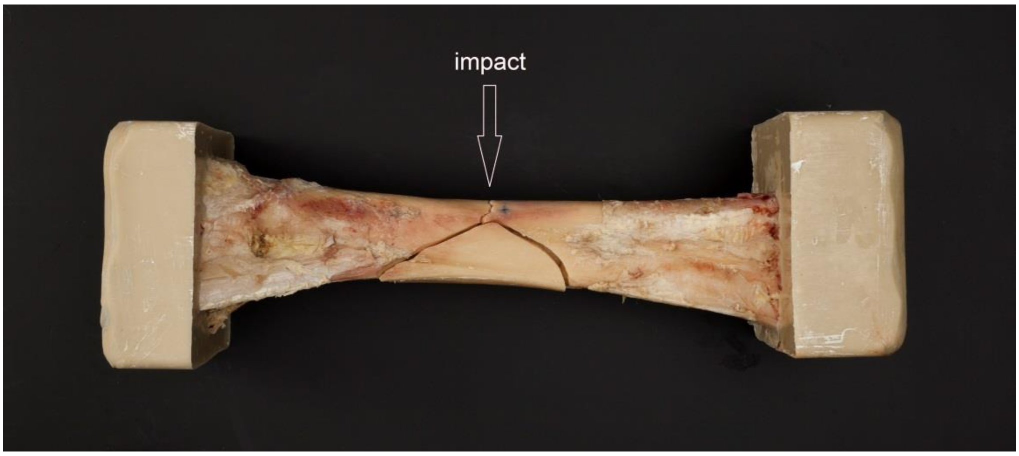

Fig. 5: Butterfly fracture of a left radius. The tip of the wedge-shaped fragment characteristically points toward the site of impact.

Impact at 12 m/s

PU: Of the 16 bones impacted by PU, four were fractured: two had a simple oblique fracture and two had a butterfly fracture. The fracture probability was 25 % with a 95 % CI [7 % to 52 %].

Hoof horn: Of the 16 bones impacted by hoof horn, two were fractured: one had a simple transverse fracture and one a simple oblique fracture. The fracture probability was 12.5 % with a 95 % CI [2 % to 38 %]. The fracture probability of bones impacted by PU and horn were largely overlapping.

The fracture probabilities from all kick experiments are summarised in Figure 6, and the fracture categories are shown in supplementary Table 1. The measured true impact velocities were on average 3.3 % and 0.84 % below the nominal values of 8 m/s and 12 m/s, respectively. Nevertheless, the comparisons between the materials are not influenced by these small and material independent deviations.

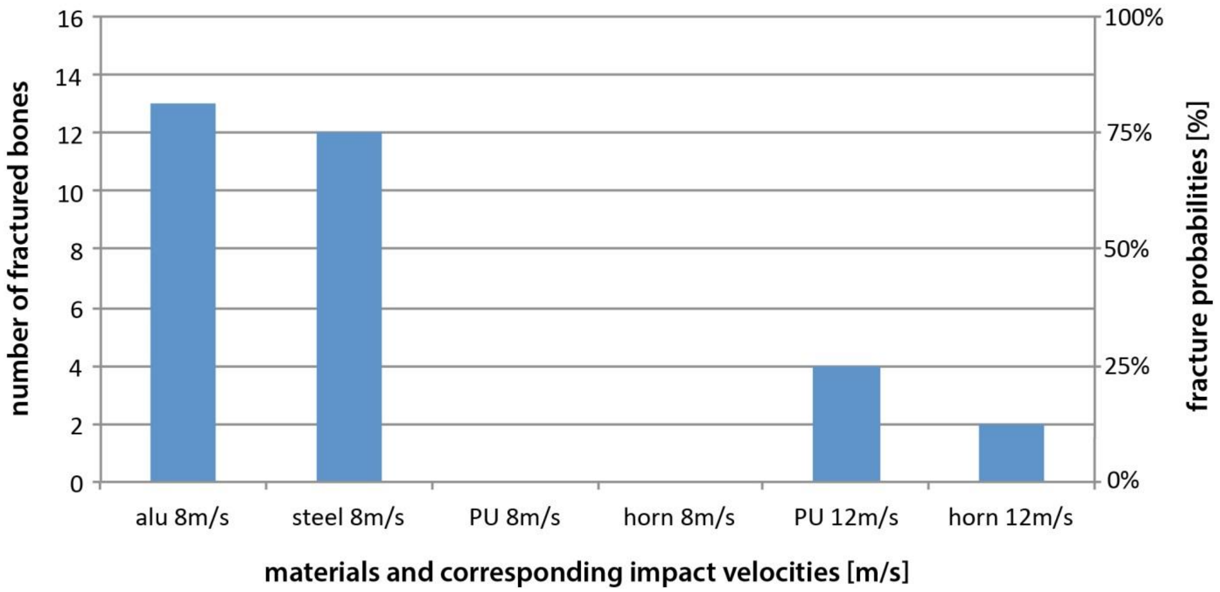

Fig. 6: Number of fractured bones and fracture probability after a simulated kick.

Physical parameters characterising the impact

Only data from experiments that did not result in fracture were analysed to eliminate the effects of individual bone strength. The descriptive statistics of the analysis are shown in Table 2.

		hard materials	**horn**	**polyurethan**
F_{peak} (kN)	**mean**	18.8	14.7	13.5
	sd	2.19	1.96	2.07
t_{impact} (ms)	**mean**	1.24	1.67	1.53
	sd	0.153	0.142	0.138

Table 2: Mean and standard deviations (sd) of the peak contact force (F_{peak}) and duration of the impact (t_{impact}) at the impact velocity of 8 m/s for the tested materials.

Peak contact force, F_{peak}

F_{peak} of hard materials was significantly larger than that of PU and hoof horn ($p < 0.001$; Tables 2 and 3). The peak contact force on the tibia was an average of 11% higher than that of the radius ($p = 0.014$; Table 3). The average peak contact forces and the predicted values based on the empirical function for a steel impactor head as calculated previously (13) are shown in Figure 7. The peak contact forces of hard materials matched the fit calculated previously (13) ($F_{peak} \approx 0.926 \cdot v_i^{1.45}$; (13)) whereas the peak contact forces of soft materials were 25 % lower. At an impact velocity of 12 m/s, the soft materials had F_{peak} values similar to those of hard materials at 8 m/s (Figure 7).

Physical parameters	**Intercept**	**Material effects**				**Bone-type (R/T) effects**	
		p-value	**Δ H - PU**	**Δ HM - PU**	**Δ HM - H**	**p-value**	**Δ T - R**
F_{peak} (kN)	12.74 [11.63 to 13.84]	< 0.001	1.13 [-0.52 to 2.77]	5.13 [3.03 to 7.24]	4.01 [1.90 to 6.11]	0.01	1.78 [0.54 to 3.01]
t_{impact} (ms)	1.566 [1.486 to 1.647]	< 0.001	0.145 [0.03 to 0.27]	-0.282 [-0.44 to -0.13]	-0.428 [-0.58 to -0.27]	0.09	-0.092 [-0.18 to -0.001]

Table 3: Results of the regression analysis assessing the effect of “material type” and “bone” on the peak contact force “F_{peak}” and the impact duration “t_{impact}”. Next to the intercept (corresponding to reference of polyurethane and radius), the p-values of “material type” and “bone” and their corresponding effect sizes (Δ), the differences, with their 95% confidence intervals are displayed. PU = Polyurethane, H = Horn, HM = ‘Hard Materials’; R = Radius, T = Tibia, Δ = difference.

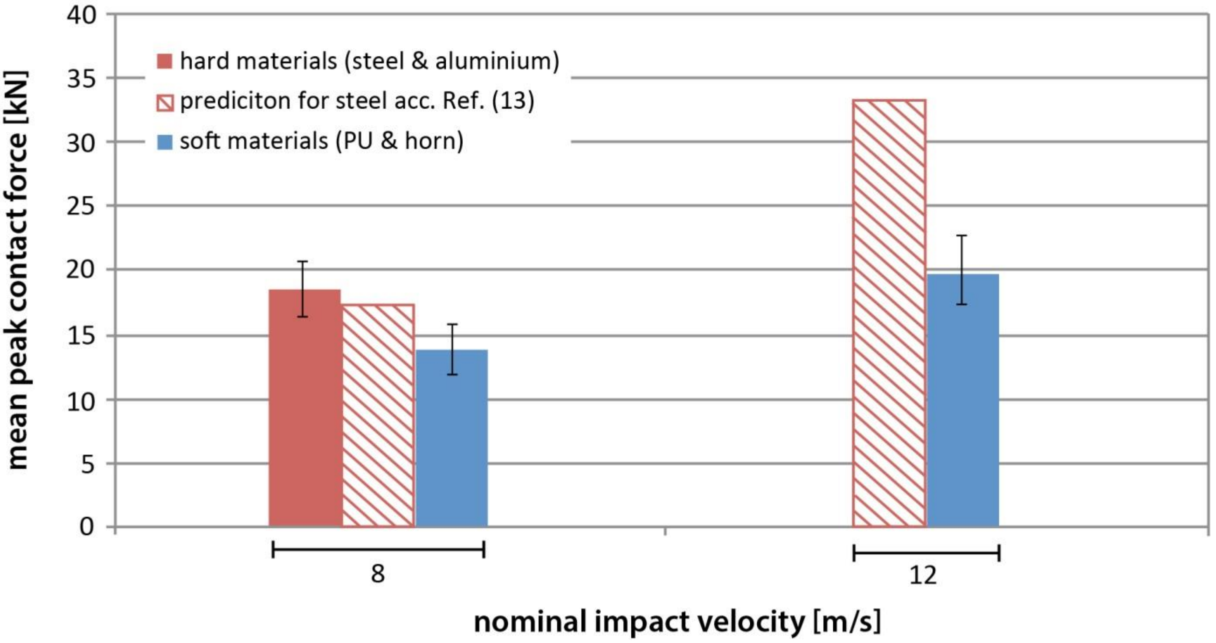

Fig. 7: Peak contact forces (F_{peak}) in case of no damage at nominal impact velocity v_i = 8 m/s and 12 m/s. The predicted values for steel have been calculated as $F_{peak} \approx 0.926 \cdot v_i^{1.45}$ (13) for the actual average impact velocities 7.73 m/s and 11.90 m/s.

Impact duration, t_{impact}

The average values of t_{impact} at 8 m/s for the different impactor heads are shown in Table 2. The hard materials had an impact duration that was 19 % and 26 % lower than that of PU and horn, respectively (both $p < 0.001$). Impact duration did not differ significantly between radius and tibia ($p = 0.094$).

There was no significant interaction between type of bone (radius/tibia) and impactor material for F_{peak} and t_{impact}.

Discussion

The probability of fracture of a long bone after a simulated kick using a steel or aluminium impactor head and a polyurethane or hoof horn impactor head at a velocity of 8 m/s (corresponding to an impact energy of E_i = 64 J) differed considerably; steel and aluminium had fracture probabilities of 75 % and 81 %, respectively, whereas the softer materials (PU and horn) did not cause detectable bone damage, neither macroscopically nor after CT examination. This in turn means that the impact energy is not the only determining factor of the fracture probability. When the impactor velocity was increased to 12 m/s (E_i = 144 J), the fracture probabilities of the softer materials were still lower (PU 25 %, horn 12.5 %) than the probabilities of the harder materials at 8 m/s. Based on these observations, the primary hypothesis that PU and horn cause significantly less damage to bone than steel or aluminium was accepted, at least under the experimental conditions used. Hence, assuming that the physical parameters of the experimental impact are comparable to those of a kick delivered by a horse, this study substantiates the empirical assumption that soft shoeing materials (PU) and unshod hoof bear a lower risk of severely injuring a horse than hard shoeing materials. Based on our observations, the common prohibition of steel or aluminium horseshoes in group housing systems can be designated as scientifically justified. The differences in fracture probability between soft and hard materials, whilst introducing the same amount of energy into the bone, can be attributed to higher compliance and damping of soft materials, which can absorb more energy and therefore have lower peak contact forces than hard materials. The peak contact force is a crucial parameter affecting the probability of a bone fracture (1, 2). The average peak contact force at 12 m/s for soft materials was, by chance, at the same level as for hard materials at 8 m/s. But the fracture probabilities were not. Hence, just as E_i, F_{peak} can’t be the sole influencing factor that affects the probability of a fracture either. Most likely, the dynamics of the impact, including the bone’s strain-rate dependent fracture resistance and dampening properties (17-19) as well as wave propagation (energy density) in the bone, play a role as well. Indeed, the impact duration, which characterises the dynamics of the impact, is considerably shorter for hard than for soft materials at the same impact velocity (Table 2). Combining steel and aluminium into a single group of hard materials was justified because of the small number of bones that were not damaged by these materials and the similarity between their properties. Furthermore, the confidence intervals of the probability of causing a fracture were similar for the two metals.

Of note, the probability of fracture in bones impacted by polyurethane and hoof horn did not differ and the physical impact parameters (F_{peak}, t_{impact}) of the two materials were almost identical. Therefore, we rejected the secondary hypothesis that a kick delivered to a long bone by an unshod hoof causes less damage than a kick from a hoof with a PU shoe.

The physical boundary conditions (velocity range, impactor weight, bone pre-load) used in this study have been shown to be representative of a real-life horse kick (13, 14). In fact, the injuries generated in the current experimental study are comparable to the clinical cases described in the literature (6) and therefore, the findings of the present study are considered clinically relevant. It could be shown, that fracture of the bone is not influenced by the boundary condition of its fixation: Fracture appeared clearly earlier than the bending wave reflexion from the bones end arrived back at the mid section (13).

Analysing the undamaged bones, the peak contact force met by tibiae was found to be significantly greater (by 11 %) than the peak contact force met by radii. This observation parallels the difference in bending stiffness of these two bone types measured in a similar, but static load case (20). This parallelism can be explained, as both reactions of the bones to an applied load, dynamic or static, are dependent on the bone material stiffness and bone geometry. In the present study, all soft tissues including the periosteum were removed from the specimens to simulate a worst-case-scenario and to facilitate comparison with results of analogous previous studies (13, 14). The attenuating effect of soft tissues plays a major role in protecting bones from serious injury and has been described in various studies (21-24).

Long-term storage of some of the bones for up to 15 years at -20 °C was a potential complicating factor in the present study. Even though the freezing process, storage at -20 to -196 °C (25, 26) or at -18 °C for up to 232 days (27, 28) and multiple thawing, testing and refreezing cycles (29) were shown to have no significant effect on the biomechanical properties and histological morphology of bones, the effect of long-term storage on bones has not been studied. Piskoty et al. used the same experimental setup described in the present study except that the bones had been frozen for no longer than eight months (13). The probability of a long-bone fracture using a steel impactor at a velocity of 8 m/s was 80 % in that study (13), which was in agreement with our findings (75 %). We therefore feel that storage at -20 °C for up to 15 years does not significantly affect the biomechanical properties of long bones, such as the resistance to an impact load. However, further studies are necessary to quantify the effect of long-term freezer storage on biomechanical properties of bones.

Conclusions and outlook

Simulation of a horse kick showed that the probability of fracture of a long bone was significantly greater with steel and aluminium impactor heads than PU and hoof horn impactor heads under experimental conditions. This strongly suggests that the shoeing system has a significant influence on the risk of injury regarding a horse kick against horses or humans. For future studies, the effect of the shoe design, e.g. geometry and structure, on the risk of injury should also be assessed, aiming for optimized shoeing systems.

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[e] Somatom Sensation Open, Siemens Medical Solutions, Zurich, Switzerland

[f] Phantom V12.1 Vision Research

[g] Matrox Design Assistant 4.0

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Anhang:

Identification of bone								**Fracture**	
No.	**Horse**	**Breed**	**Gender**	**Age**	**Source**	**Kick velocity [m/s]**	**Impactor Material**	**Main type**	**Subtype**
1	I	WB	f	9	TR	8	A	3	5
2	I	WB	f	9	TL	8	S	1	
3	I	WB	f	9	RR	12	PU	1	
4	I	WB	f	9	RL	12	H	1	
5	II	WB	m	18	TR	8	A	3	2
6	II	WB	m	18	RL	8	S	3	2
7	II	WB	m	18	RR	12	PU	3	2
8	II	WB	m	18	TL	12	H	1	
9	III	XX	m	13	RL	8	A	3	4
10	III	XX	m	13	RR	8	S	1	
11	III	XX	m	13	TR	12	PU	3	4
12	III	XX	m	13	TL	12	H	3	1
13	IV	WB	m	4	RR	8	A	1	
14	IV	WB	m	4	RL	8	S	1	
15	IV	WB	m	4	TL	12	PU	1	
16	IV	WB	m	4	TR	12	H	1	
17	V	WB	m	13	TR	8	A	1	
18	V	WB	m	13	TL	8	S	2	
19	V	WB	m	13	RL	12	PU	1	

20	V	WB	m	13	RR	12	H	1	
21	VI	WB	m	5	RL	8	A	2	
22	VI	WB	m	5	RR	8	S	2	
23	VI	WB	m	5	TL	12	PU	1	
24	VI	WB	m	5	TR	12	H	1	
25	VII	Quarter	m	12	TR	8	A	1	
26	VII	Quarter	m	12	TL	8	S	1	
27	VII	Quarter	m	12	RL	12	PU	1	
28	VII	Quarter	m	12	RR	12	H	1	
29	VIII	WB	m	14	TL	8	A	2	
30	VIII	WB	m	14	TR	8	S	3	2
31	VIII	WB	m	14	RR	12	PU	1	
32	VIII	WB	m	14	RL	12	H	1	
33	IX	WB	f	18	RR	8	A	3	2
34	IX	WB	f	18	TR	8	S	3	2
35	IX	WB	f	18	TL	12	PU	1	
36	IX	WB	f	18	RL	12	H	1	
37	X	WB	m	26	RL	8	A	3	5
38	X	WB	m	26	RR	8	S	3	2
39	X	WB	m	26	TR	12	PU	3	4
40	X	WB	m	26	TL	12	H	1	
41	XI	WB	f	11	TL	8	A	3	4
42	XI	WB	f	11	TR	8	S	2	
43	XI	WB	f	11	RR	12	PU	1	
44	XI	WB	f	11	RL	12	H	1	

45	XII	FM	m	22	RL	8	A	3	3
46	XII	FM	m	22	RR	8	S	3	1
47	XII	FM	m	22	TR	12	PU	3	2
48	XII	FM	m	22	TL	12	H	1	
49	XIII	Tinker	m	21	RL	8	A	2	
50	XIII	Tinker	m	21	TR	8	S	3	2
51	XIII	Tinker	m	21	RR	12	PU	1	
52	XIV	WB	m	14	TL	8	A	3	4
53	XIV	WB	m	14	RR	12	PU	1	
54	XIV	WB	m	14	RL	12	H	1	
55	XV	XX	m	11	TR	8	A	3	2
56	XV	XX	m	11	TL	8	S	3	5
57	XV	XX	m	11	RR	12	PU	1	
58	XVI	FM	m	15	RR	8	A	3	1
59	XVI	FM	m	15	TR	8	S	3	2
60	XVI	FM	m	15	TL	12	PU	1	
61	XVII	WB	m	21	RL	8	S	3	3
62	XVII	WB	m	21	TR	12	PU	1	
63	XVII	WB	m	21	TL	12	H	3	2
64	XVIII	XX	m	16	RL	12	H	1	

Supplementary Table 1: Frequency of incomplete and complete fractures after a simulated kick.

Breed: WB = warmblood, XX = thoroughbred, FM = Freiberger, m = gelding/stallion, f = mare, RR = right radius, RL = left radius, TR = right tibia, TL = left Tibia, S = Steel, A = Aluminium, PU = Polyurethane, H = Horn.

Main type: 1 = no damage, 2 = fissure, 3 = complete fracture

Subtype: 1 = transverse fracture, 2 = oblique fracture, 3 = longitudinal fracture, 4 = butterfly fracture, 5 = comminuted fracture